Changing stride frequency alters average joint power and power distributions during ground contact and leg swing in running.

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Abstract

Purpose. Runners naturally adopt a stride frequency closely corresponding with the stride frequency that minimizes energy consumption. While the concept of self-optimization is well recognized, we lack mechanistic insight in the association between stride frequency and energy consumption. Altering stride frequency affects lower extremity joint power, however these alterations are different between joints, possibly with counteracting effects on the energy consumption during ground contact and swing. Here, we investigated the effects of changing stride frequency from a joint level perspective.

Methods. 17 experienced runners performed six running trials at five different stride frequencies (preferred stride frequency (PSF) twice, PSF ± 8%, PSF ± 15%) at 12 km/h. During each trial, we measured metabolic energy consumption and muscle activation, and collected kinematic and kinetic data which allowed us to calculate average positive joint power using inverse dynamics.

Results. With decreasing stride frequency, average positive ankle and knee power during ground contact increased ($p < 0.01$) while average positive hip power during leg swing decreased ($p < 0.01$). Average soleus muscle activation during ground contact also decreased with increasing stride frequency ($p < 0.01$). In addition, the relative contribution of positive ankle power to the total positive joint power during ground contact decreased ($p = 0.01$) with decreasing stride frequency whereas the relative contribution of the hip during the full stride increased ($p < 0.01$) with increasing stride frequency.

Conclusion. Our results provide evidence for the hypothesis that the optimal stride frequency represents a trade-off between minimizing the energy consumption during ground contact, associated with higher stride frequencies, without excessively increasing the cost of leg swing or reducing the time available to produce the necessary forces.

Keywords: locomotion, self-optimization, step length, joint work

Introduction

Runners naturally adopt running kinematics associated with minimal energy consumption, often referred to as self-optimization (1). Stride frequency is one of the variables naturally selected and – although highly variable between runners – the preferred stride frequency (PSF) closely matches the metabolically optimal stride frequency, i.e., the stride frequency that results in the lowest energy consumption (1–3). Running with a stride frequency lower or higher than the optimal frequency increases whole-body metabolic energy consumption per distance travelled resulting in a U-shaped frequency – metabolic cost curve (1–3).

While several studies have tried to explain the U-shaped relationship between metabolic energy consumption and stride frequency, no clear mechanism has been identified yet. Cavagna and colleagues (1988) (4) found that hopping or fast running animals adopt a stride frequency slower than the symmetrically bouncing frequency, i.e., the frequency at which the timespan of the vertical force exceeding body weight equals the time where vertical force is lower than body weight. While this symmetrical bouncing frequency minimizes the external work (work done on the body's center of mass, COM), by selecting a lower stride frequency the animals avoid an increase in internal work (work required to accelerate and decelerate the limbs relative to the body's COM) associated with higher frequencies. In humans, the optimal stride frequency is similarly proposed to represent a trade-off between external and internal work as to minimize the total mechanical work (5). However, this method to calculate total mechanical work based on the sum of external and internal work largely underestimates the total muscular work (6). A better approach is to calculate individual joint average powers based on inverse dynamics and sum all these average joint powers to obtain total lower limb average powers (7,8).

While many studies have focused on the energetically expensive ground contact phase, swing phase cost is likely to be sensitive to stride frequency as well. Running can be divided into a ground contact and leg swing phase. During ground contact the majority of positive power is performed around the ankle joint (9) and the leg muscles produce force to support body weight and propel the body forward (10). While the ground contact phase is energetically the most expensive phase (11), during leg swing some - mostly hip - muscles are active to swing the leg forward and consume energy. Several studies have estimated the relative contribution of leg swing to the net metabolic cost of running ranging from 7 to 26% (11–13). Moreover, Doke et al. (2005) (14) demonstrated that the cost of swinging an isolated leg sharply increases with swing frequency ($\dot{E}_{net} \sim frequency^4$) and that positive mechanical work around the hip strongly correlates with the cube of leg swing frequency ($R^2 = 0.93$). These results suggest that increasing the stride frequency beyond the optimal frequency may substantially increase the energy consumption during leg swing and as such increases the overall rate of metabolic energy consumption. Yet, while this mechanism can explain the increased metabolic cost when increasing stride frequency above the optimal frequency, it cannot explain the increase in metabolic cost with decreasing stride frequency below the optimal frequency. Studies connecting the legs with a spring (i.e. exotendon), finding reductions in energy consumption of 6 to 8 % (15,16), highlight the interaction between joints while running at certain stride frequency. Simpson et al. (2019) (16) demonstrated that by reducing swing work, through the exotendon, runners adopt a higher stride frequency reducing joint powers around the ankle and knee joint during ground contact and leg swing, emphasizing the complex interaction between lower extremity joints. Hence, investigating the effect of changing stride frequency on lower limb positive joint power and subdividing these joint powers into ground contact and leg swing may enhance our understanding on the mechanisms determining the optimal stride frequency.

Here we hypothesize that with increasing stride frequency the average positive hip joint power during leg swing increases while the average positive ankle joint power during ground contact decreases. In addition, we expect that increasing stride frequency would redistribute positive joint power from the ankle joint towards the more proximal hip joint. To test this hypothesis, we used an inverse dynamic approach to calculate individual joint moments and joint powers. In addition, we measured muscle activity of the ankle plantar flexors as changes in positive power may reflect in different muscle activations providing more direct evidence of altered energy consumption.

Materials and methods

Participants. Seventeen (body mass: 69.1 ± 7.7 kg; height: 1.79 ± 0.09 m; age: 23.7 ± 3.8 y; 13 male; 4 female) injury free subjects gave written informed consent, approved by the local ethical committee, and participated in this study. All subjects were capable of running 5 km under 20 minutes (16’13 ± 1’33 [range: 13’19 – 19’00]) and ran at least 30 km/week.

Experimental setup. Subjects performed a warm-up on a force measuring treadmill (Motekforce Link, Amsterdam, The Netherlands) at a self-selected speed for a period of at least 5 minutes. Next, the treadmill velocity was set at 12 km/h and after several minutes the preferred stride frequency of each participant was determined by counting the number of strides taken during a one-minute time interval. This provided the participants with ample treadmill running exposure before we quantified preferred stride frequency. The average of the three minutes was considered as the preferred stride frequency. Participants ran six 5-minute trials at a constant speed of 12 km/h with each trial adopting a different stride frequency (PSF, PSF ± 8% and PSF ± 15%) enforced by a metronome and verified using the ground reaction force data. Subjects had 5 minutes of rest between trials. We chose the running speed of 12 km/h based on Hoogkamer et al. (2016) (17) who had participants of similar fitness running at 3.5 m/s (12.6 km/h) with added mass to their shoes, demonstrated to increase metabolic energy consumption.

During the first and last trial, subjects ran at their preferred stride frequency, the stride frequencies for the four other trials were randomized. During each trial, ground reaction forces, marker trajectories and whole-body metabolic energy consumption data were collected.

Metabolic energy consumption. We measured whole-body metabolic energy consumption using indirect calorimetry (Cosmed K5, Cosmed srl, Rome, Italy). Prior to testing the flow turbine, oxygen and carbon dioxide analyzers were calibrated according to the manufacturer's instructions. Rates of oxygen consumption and carbon dioxide production were collected and averaged over the last 90 seconds. We computed whole-body metabolic energy consumption (in Watts) using the Brockway equation (18) and normalized energy consumption to subject's body mass. To allow for reliable calculation of aerobic metabolic energy consumption, subjects should be running at submaximal intensity which we verified based on the respiratory exchange ratio. One subject's respiratory exchange ratio exceeded 1.0, indicating that the subject was no longer running at submaximal intensity, and that subject was discarded for further analysis. Since the PSF condition was measured during two trials, we took the average whole-body metabolic energy consumption of both trials, except for one subject where we had issues during the first measurement and therefore only used the last trial.

Kinetics and kinematics. Thirteen infrared motion capturing cameras (Vicon, Oxford Metrics, Oxford, UK) recorded the motion of 48 reflective makers, including four cluster markers on the thigh and shank, at a sampling frequency of 200 Hz. Ground reaction force (GRF) data, measured at 1000 Hz, and marker trajectory data were low-pass filtered with a cut-off frequency of 20 Hz. We used the filtered GRF data to determine ground contact, adopting a 30 N threshold, and to calculate the actual stride frequency and duty factor (ground contact time divided by stride time).

A marker labeled static trial (Nexus 2.4, Oxford Metrics, UK) was used to scale the Hamner musculoskeletal model (19) according to the subject's dimensions in OpenSim 3.3 (OpenSim, Stanford, CA, USA). Based on the dynamic marker trajectory data, joint angles were computed using a Kalman Smoothing algorithm (20). Next, we conducted an inverse dynamic analysis in OpenSim which, based on the dynamic equations of motion, calculates joint torques. Briefly, joint torques are computed using the joint angles, ground reaction forces, segment masses, segment moments of inertia and segment (angular) accelerations. Joint torques were low-pass filtered using a recursive fourth-order Butterworth filter with a cut-off frequency of 20 Hz and multiplied by the respective joint angular velocity to compute joint power at the hip, knee and ankle. After normalizing joint power to the subject's body mass, we calculated positive joint work by integrating positive joint power with respect to time. To allow for comparison between conditions, we divided positive joint work during a full stride by stride time to calculate average positive joint power. Accordingly, average positive joint power during ground contact and during swing were computed as the positive joint work during ground contact or swing and divided by stride time. Finally, to calculate the relative contribution of each joint to the total positive average joint power during the full stride we divided the average joint power of each joint by the sum of the average positive power of the hip, knee and ankle. Similarly, to compute the relative contribution of each joint during ground contact, positive average joint power during ground contact was divided by the sum of positive average joint power of the hip, knee and ankle during ground contact only.

Electromyography. We measured the muscle activity of the major ankle plantar flexor muscles (gastrocnemius medialis, GM; gastrocnemius lateralis, GL; and soleus, SOL) and ankle dorsiflexor muscle (tibialis anterior, TA) through surface electromyography (Zerowire, CA, US) with a sampling frequency of 1000 Hz. Before placing the bipolar EMG electrodes (Ag/AgCl electrodes, 10 mm recording diameter, Ambu), we shaved and cleaned the skin with alcohol gel. EMG electrodes were placed on the muscle belly of the GM, GL and TA parallel to the muscle fibers with an inter-electrode distance of 2 cm. The SOL electrodes were placed at 2/3 of the line between lateral condyle of the femur and the lateral malleolus, parallel with the muscle fibers and 2 cm apart. The raw EMG signal was first band pass filtered (20-400 Hz), rectified and low pass filtered (20 Hz). To compare muscle activation during ground contact between stride frequency conditions, we calculated the time-integral of the EMG signal during ground contact. We normalized the integrated EMG signal to the peak amplitude of the EMG signal, adopting a 10 ms moving average window, of each muscle across all conditions and for every participant. Finally, to calculate the average activation per unit time during ground contact, we divided the normalized and integrated EMG signal of each muscle during ground contact by the stride time. Due to issues with the EMG equipment, we did not record muscle activation for one participant. We visually inspected all EMG signals for each participant and for every condition. Five corrupted EMG files were discarded (one participant's GM, one participant's GL, one participant's TA and two participants' SOL).

Statistics. All data are presented as mean ± standard deviation. Data were first tested for normality and sphericity using the Shapiro-Wilk test and Mauchly's test, respectively. For normally distributed data, we conducted a repeated measures ANOVA to test for significant differences between stride frequency conditions. If the assumption of sphericity was violated, we performed the Greenhouse-Geisser correction. When the data were not normally distributed, we executed the non-parametric Friedman test. If a significant main effect was found, we used the Bonferroni correction for post-hoc testing to identify which conditions were significantly different from the PSF condition. We also calculated partial eta squared (η^2) as a measure for effect size for the repeated measure ANOVA where $\eta^2 \geq 0.26$ is considered as a large effect. If repeated measure ANOVA could not be performed due to violations against normality, we calculated Kendall's W where $0.3 \leq W < 0.5$ indicates a moderate effect and $W > 0.5$ a strong effect. Statistical significance was set at $p < 0.05$. An a priori power calculation (G*Power version 3.1) indicated that, to detect significant changes in average positive ankle or knee joint power during ground contact (ES = 0.91 (9) and power = 0.8), we needed 15 participants.

Table 1. Kinematic data as mean ± SD for each stride frequency condition (N = 16).

	-15%	**-8%**	**PSF**	**+8%**	**+15%**
Stride frequency (strides/min)	73.0 ± 2.8*	78.2 ± 3.0*	84.0 ± 3.2	90.1 ± 3.5*	95.7 ± 4.0*
Step length (cm)	137 ± 5*	128 ± 5*	119 ± 5	111 ± 4*	105 ± 5*
Ground contact time (ms)	229 ± 22	227 ±18	222 ± 17	213 ± 18	206 ± 17*
Duty factor (%)	27.9 ± 2.4*	29.5 ± 2.3*	31.1 ± 2.3	32.0 ± 2.6*	32.9 ± 2.4*

** represents significantly different from the self-preferred stride frequency condition (PSF).*

Results

Actual stride frequencies were substantially different from the preferred frequency, on average stride frequencies were -13.1%, -7.0%, +7.2% and +14.0% different from PSF (Table 1). With increasing stride frequency ground contact time significantly decreased ($p < 0.001$; W = 0.75) while duty factor significantly increased ($p < 0.001$; $\eta^2 = 0.84$). Whole-body metabolic energy consumption followed a U-shaped curve, with the lowest energy consumption corresponding with the preferred frequency ($p < 0.001$; $\eta^2 = 0.66$; Figure 1). Post-hoc analysis revealed that energy consumption at all except the +8% stride frequency condition was significantly different from PSF. When looking at the individual data, 12 out of the 16 subjects demonstrated the lowest energy consumption at PSF. For the other four subjects, three of them showed minimal energy consumption when running at PSF +8% and one while running at PSF -8%. Yet, the difference in energy consumption between the frequency associated with minimal energy consumption and PSF were relatively small for those subjects, within the typical measurement error for metabolic energy consumption (21).

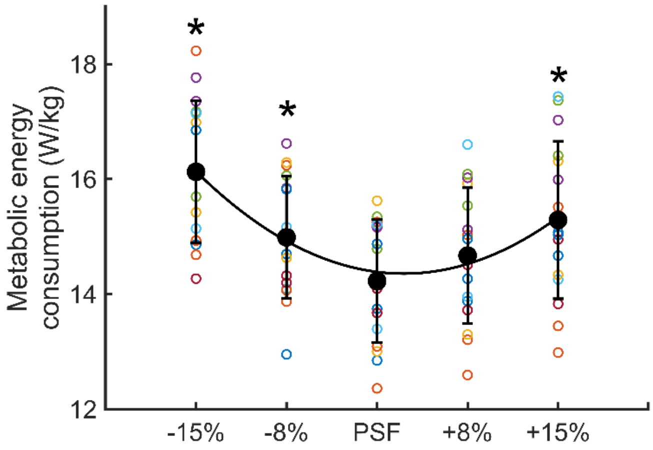

*Figure 1. Metabolic energy consumption across the five stride frequency conditions. Solid black dots represent means and error bars SD. Open circles are the individual data. *significantly different from PSF.*

Average positive ankle and knee joint power during ground contact decreased with increasing stride frequency ($p < 0.001$; ankle: $\eta^2 = 0.65$; knee: $\eta^2 = 0.75$; Figure 2). Post-hoc analysis demonstrated that average positive ankle joint power was only significantly different from PSF at lower stride frequencies, whereas for the knee joint significant differences in average positive joint power were found between PSF and PSF ± 15%. Running at the lowest stride frequency increased average ankle joint positive power by 13% compared to PSF. In line with the increase in average positive ankle joint power, the average soleus muscle activation during ground contact significantly increased when stride frequency decreased ($p < 0.01$; $\eta^2 = 0.28$; Figure 3). Post-hoc analysis revealed that average soleus activation during ground contact when running at PSF +15% was significantly lower compared to PSF. We did not find any significant difference in average muscle activation during ground contact across stride frequency conditions for the other muscles.

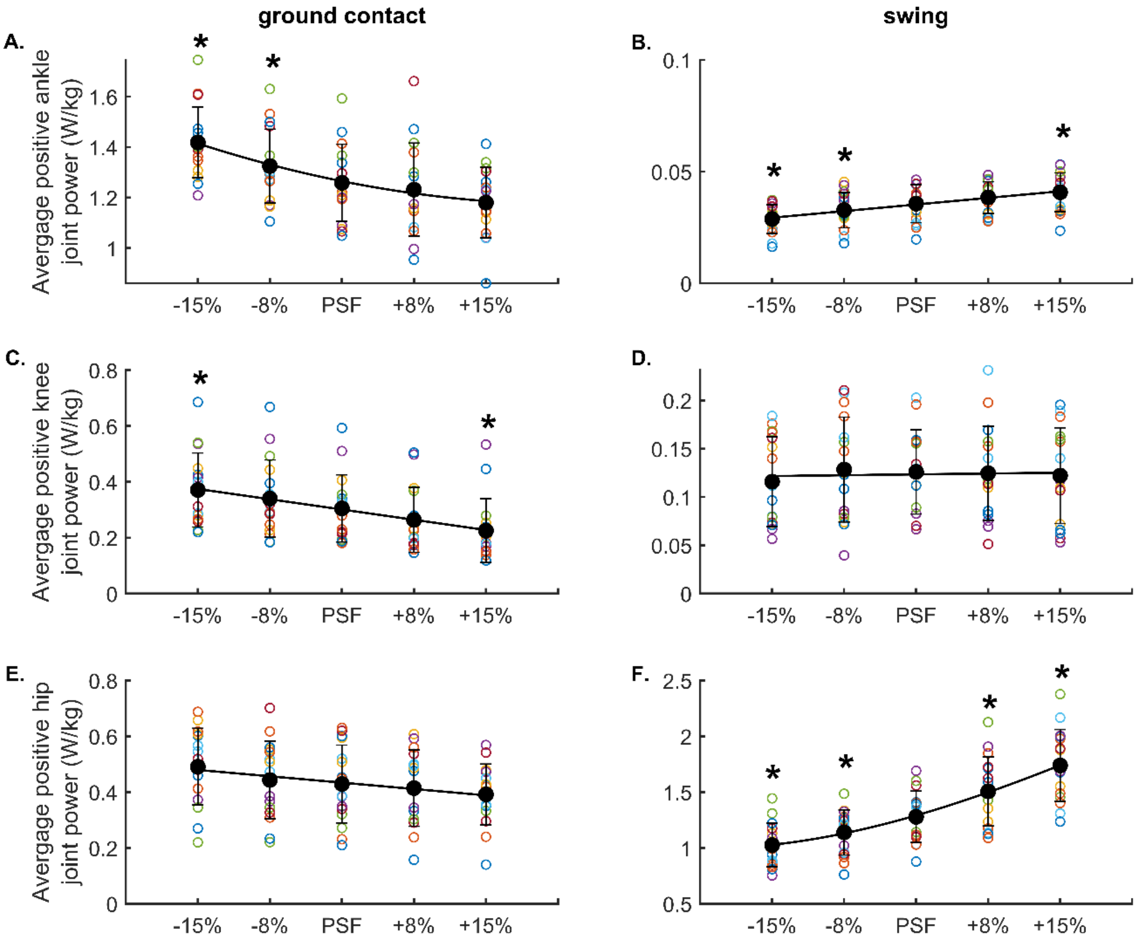

*Figure 2. Average positive ankle (A,B), knee (C,D) and hip (E,F) power during ground contact (A,C,E) and leg swing (B, D, F) during the five stride frequency conditions. Solid black dots are the means and error bars the SD. Open circles represent the individual data. *significantly different from PSF.*

During leg swing, average positive hip joint power strongly increased with increasing stride frequency ($p < 0.001$; $\eta^2 = 0.81$). Post-hoc analysis revealed significant differences in average positive hip joint power between all condition and PSF. At the lowest stride frequency, hip average positive power was reduced by 20% whereas at the highest stride frequency it increased by 36%. As such, the relative contribution of hip average positive power during the full stride increased with increasing stride frequency ($p < 0.001$; $\eta^2 = 0.87$; Figure 4), while the contribution of ankle and knee reduced ($p < 0.001$; ankle $\eta^2 = 0.76$, knee $\eta^2 = 0.75$). In contrast, during ground contact the relative contribution of the ankle joint slightly increased with increasing stride frequency ($p = 0.01$; $W = 0.24$) whereas the contribution of the knee joint decreased ($p < 0.001$; W = 0.53). We found no difference in relative contribution of the hip joint during ground contact.

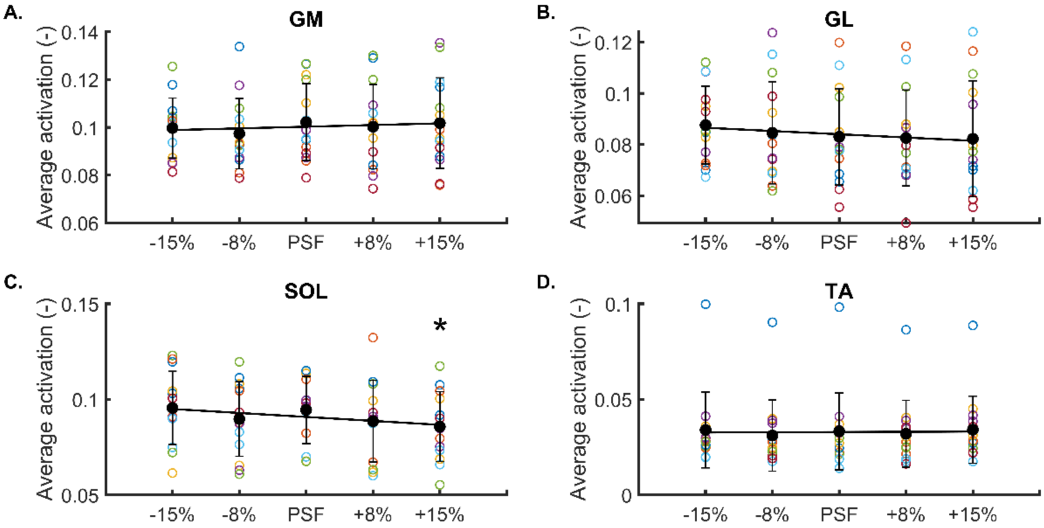

*Figure 3. Average muscle activation during the ground contact phase of running of the gastrocnemius medialis (A.; N = 14), gastrocnemius lateralis (B.; N = 14), soleus (C.; N = 13) and tibialis anterior (D.; N = 14) across five stride frequency conditions. Solid black dots are the means and error bars the SD. Open circles represent the individual data. *significantly different from PSF.*

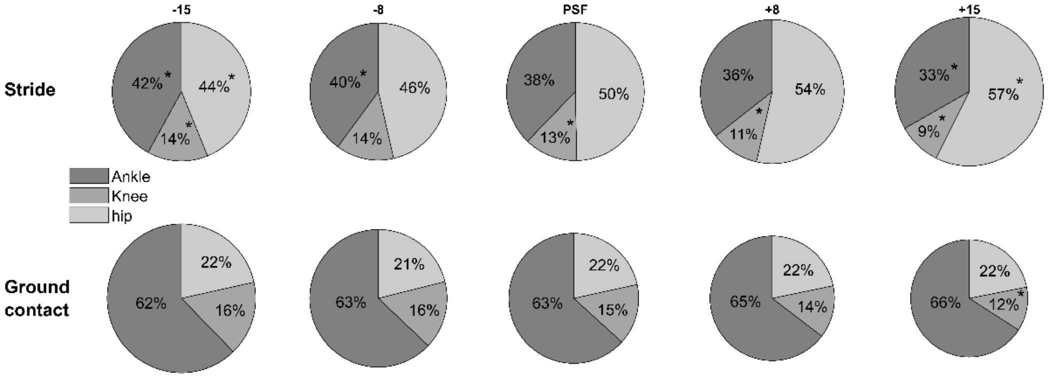

*Figure 4. Relative contribution of the ankle (dark grey), knee (grey) and hip (light grey) to the total average positive power during the full stride (top) and ground contact only (bottom). The radius of each pie chart is scaled based on the total positive power in each condition. *significantly different from PSF.*

Discussion

In this study, we investigated the effect of altering stride frequency on average positive joint power and positive joint power distribution during the ground contact and swing phase of running. We accept our first hypothesis that increasing stride frequency decreases average positive ankle power during ground contact but increases average positive hip power during leg swing. With increasing stride frequency, the sharp increase in hip power during leg swing implied that the majority of positive power during a full stride was provided by the hip. In contrast, increasing stride frequency also redistributed average positive joint power from the knee towards the ankle during ground contact. Our results suggest that the mechanisms inducing the increase in metabolic energy consumption when adopting a stride frequency higher or lower than the optimal frequency are different.

At stride frequencies below the PSF, the large increase in positive ankle power and the small decrease in positive hip power with decreasing stride frequency might explain why the net result is an increase in energy consumption. In addition, average positive knee joint power during ground contact increases with decreasing stride frequency. These results support previous research demonstrating that decreasing stride frequency increases positive joint work during ground contact for both the ankle and knee joint (22) and reduces braking forces impulses (23). The ground contact phase in running is energetically the most expensive phase (11), with the ankle joint providing most of the positive power during ground contact (Figure 4). Previous research already estimated that the Triceps Surae muscle, the major plantar flexor muscle, consumes between 20 and 40% of the total energy during running at the preferred stride frequency (24,25). Although increases in average positive ankle joint power may be partly provided by more elastic energy storage and return it will likely increase muscle force or work. This hypothesis is further supported by an increase in average soleus activation during ground contact with decreasing stride frequency (Figure 3). Hence, the increase in average positive ankle joint power, associated with decreasing stride frequency, may increase the energy consumed by the Triceps Surae during ground contact.

Next to the large increase in positive ankle power with decreasing stride frequency, there is also a relatively large increase in positive knee power which might explain an increase in whole-body metabolic energy consumption. In absolute terms, the increase in positive ankle joint power with decreasing stride frequency is much larger than the increase in positive knee joint power, yet there is a change in the relative distribution of joint power during ground contact (Figure 4). At lower stride frequencies, the relative contribution of the knee increases while the contribution of the ankle decreases. The Triceps Surae muscle-tendon unit spanning the ankle joint exhibits a morphology allowing for slow muscle fiber contraction velocities since most of the length changes in the muscle-tendon unit is taken up by the long, compliant in series connected elastic element (26–29). In contrast, the more proximal knee and hip muscles lack those long, compliant series elastic elements and most of the length changes in the muscle-tendon units are provided by the muscles. Therefore, the hip and knee are suggested to be metabolically less efficient than the ankle and as such, the increase in positive knee power may come with a relatively high metabolic cost. While future studies should further look into and confirm whether different stride frequencies also result in altered muscle dynamics, it adds to the idea that decreasing stride frequency will make the energy costly ground contact phase even more expensive.

At stride frequencies above the PSF, the small decrease in positive ankle power and the large increase in positive hip power with increasing stride frequency might explain why the net result is an increase in energy consumption. Previously, the cost of leg swing has often been neglected. However, several studies already estimated a substantial metabolic energy cost for swinging the legs (11–13) and previous research demonstrated that increasing stride frequency leads to increased maximal hip flexor moment during swing (23). Moreover, adding mass to the leg or foot alters the inertial properties of the leg and increases metabolic energy consumption during running (17,30,31). The more distal the mass is added, the greater the increase in energy consumption (30,31). Doke et al. (2005) (14) revealed that the cost of an isolated leg swing increases with the fourth power of swing frequency. Based on this, increasing stride frequency from 84 strides/min to 95.7 strides/min would increase the energy consumption of leg swing by

1 W/kg more than the energy consumption decrease by reducing the stride frequency with a similar amount (from 84 strides/min to 73 strides/min). While the cost of leg swing might be rather small when running at the preferred stride frequency (i.e. 7-26% of the total metabolic energy consumption (11–13)), our results indicate that the cost of leg swing may become a more substantial energetic cost when increasing stride frequency beyond the optimal frequency.

Furthermore, our results demonstrate that runners adapt their stride kinetics (i.e. duty factor) when changing stride frequency, illustrating that the interaction between increasing stride frequency and metabolic energy consumption is more complex than just average positive hip joint power. With increasing stride frequency, runners adopt a greater duty factor while running (Table 1). As such, the relative decrease in ground contact time is smaller than the actual reduction in stride time, indicating that runners alter their kinetics to prioritize time on the ground over swing time. Metabolic energy consumption during running is proposed to be inversely proportional to ground contact time (32,33). Hence, although the positive average ankle power during ground contact tends to decrease (yet not significantly different from PSF) when stride frequency increases, energy consumption during ground contact may only slightly reduce due to the shorter time available on the ground. Similarly, Doke and Kuo (2007) (34) established that the increase in metabolic cost with increasing leg swing frequency is not only determined by an increase in mechanical work but also due to a reduction in time to produce the necessary force, i.e., rate of force development. A greater rate of force development induces fast muscle activation and deactivation associated with a more energy expensive calcium pumping (34) and possibly induces the activation of less economical muscle fibers (32,35). Hence, the shorter time to produce the necessary force to swing the leg forward will increase the cost of leg swing more than what would have been expected based on average positive hip power only (34).

Some of the limitations of the study are that our participants were trained runners, running at least 30 km/week on average. Since the PSF of trained runners more closely matches their optimal stride frequency than the PSF of novice runners (36), not all results may be extrapolated to novice runners. Next, we calculated average positive joint power and used this positive power to explain altered metabolic energy consumption. While we normalized the power to time, the method is still subject to redundancy issues. The inverse approach calculates net joint powers which slightly underestimates total positive power due to antagonist muscle co-contraction (6). The muscle redundancy issues also imply that assumptions are made regarding individual muscle contractile and tendon behavior. Calculated positive power will not always represent actual muscle power due to passive in series connected elastic tissues performing most of the work, while muscles primarily produce force (35). Especially around the ankle joint where the Triceps Surae contracts almost isometrically (29), producing little work, making the muscle-tendon units around the ankle more efficient for power production (37). Future studies should use in vivo ultrasound to investigate whether muscle-tendon dynamics change with altered stride frequencies. Simulation-based studies can estimate individual muscle energy consumption and provide more insights about how lower leg muscle energy consumption is altered when running at different stride frequencies. Lastly, we only collected muscle activity data of the triceps surae muscles because, initially, we were most interested in these muscles. Yet, our results demonstrate that muscle associated with leg swing (i.e. iliopsoas, iliacus, rectus femoris, …) do play an important role and therefore future studies may want to collect muscle activity data of those muscles.

In conclusion, we found that increasing stride frequency reduces average positive ankle power during ground contact while it more than proportionally increases average positive hip power during leg swing. Our results further build on the hypothesis that the optimal stride frequency represents a trade-off between minimizing the ground contact cost, here estimated by positive ankle joint power during ground contact, and minimizing the swing cost, estimated as hip joint power during leg swing, without substantially reducing the time to produce the necessary force. Additionally, running with an increased stride frequency is often recommended as a simple strategy to reduce knee joint loading (22,38), yet our results demonstrate that, from a performance point of view, it may not be the most appropriate strategy.

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